Phase Diversity for Speckle Reduction

Ricardo G. Dantas[a,b*], Sidney Leeman[b], Eduardo T. Costa[a], Joie P. Jones[c], Eduardo J. V. Oliveira[a]

[a]UNICAMP – Departamento de Engenharia Biomédica (DEB/FEEC) – Centro de Engenharia Biomédica (CEB), Caixa Postal 6040, Campinas/SP, 13.084-971, Brazil
[b]King's College London, Strand, London, WC2R 2LS, UK
[c]Dept. of Radiological Sciences, University of California Irvine, Irvine, CA, USA 92697-5000

ABSTRACT

B-mode ultrasound images are characterised by speckle artefact, which results from interference effects between returning echoes, and may make the interpretation of images difficult. Consequently, many methods have been developed to reduce this problematic feature. One widely used method, popular in both medical and non-destructive-testing applications, is a 1D method known as Split Spectrum Processing (SSP), or also as Frequency Diversity. Alhough this method was designed for speckle reduction applications, the final image experiences a resultant loss of resolution, impinging a trade-off between speckle reduction and resolution loss. In order to overcome this problem, we have developed a new method that is an extension of SSP to 2D data using directive filters, called Split Phase Processing (SPP). Instead of using 1D narrow band-pass filters as in the SSP method, we use 2D directive filters to split the RF ultrasound image in a set of wide band images with different phases. The use of such filters substantially avoids the resolution loss usually associated with SSP for speckle reduction, because they effectively have the same bandwidth as the original image. It is concluded that the Split Phase Processing, as introduced here, provides a significant improvement over the conventional Split Spectrum Processing.

Keywords: Speckle reduction, ultrasound imaging, split spectrum processing, split phase processing.

1. INTRODUCTION

B-mode ultrasound images, as widely used for medical scanning, are characterised by the speckle artefact which results from interference effects between returning echoes, and which may make the interpretation of images difficult. The speckle artefact is responsible for introducing false 'worm-like' structures whose resolution is beyond the ultrasound system capabilities, masking the true interfaces (Burckhardt, 1978). Consequently, many methods have been developed to reduce this problematic feature, some necessitating multi-image input, with others requiring only a single image. Split Spectrum Processing (SSP), also known as frequency diversity, which is a single image technique, and angular-, or spatial-, compounding (a multi-image method) are two well-known methods. SSP is traditionally performed in one dimension (1D), where each individual radio-frequency (RF) A-line is filtered by a set of relatively narrow band-pass filters and the outputs are compounded to generate a speckle-reduced envelope that consequently exhibits a significant resolution loss (Shankar and Newhouse, 1985). The angular compounding method (Li and Odonnell, 1994) does not suffer from this problem but requires that the analysed structure be imaged from a number of different scanning directions - a severe practical limitation for medical imaging in particular.

The method proposed here may be seen as remedying the drawbacks of the two above techniques: it essentially maintains image resolution but requires only a single image as input. We have called the method 'Split Phase Processing' (SPP) because it processes image data by using two dimensional (2D) directive filters in Fourier space to split the RF ultrasound image into a set of relatively wide-band 'pre-images' with the same spectral amplitude components but with different (Fourier) phase functions. The 2D filters are chosen to have a bandwidth close to that of the original image, thereby essentially preserving the input resolution.

Results may be seen to be very encouraging indeed: resolution loss is minimal and not apparent on qualitative visual inspection. The other advantage is that, unlike the angular compounding method, SPP requires a single input image, avoiding practical image acquirement limitations. The method's effectiveness was established by using quantitative

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indices, such as signal to noise ratio (SNR) and contrast to noise ratio (CNR). The potential of the technique is apparent qualitatively as well as quantitatively. With simulated images, the possibility of constructing an 'ideal' speckle-free image exists (Gehlbach, 1983), and this is compared with the speckle reduction achievable with SPP.

It is concluded that Split Phase Processing, as a practicable single image technique, provides a significant improvement over conventional Split Spectrum Processing by reducing the speckle artefact interference and preserving edges details, regarded to its minimum resolution loss effect.

2. MATERIAL AND METHODS

The amplitude and phase of the pulse-echo lines which make up the final (envelope-detected) B-mode image are measurable and presumed known, as for the SSP technique. SSP is a widely used single image technique originally developed for radar imaging systems. After single line data are stored, the 1D-backscattered signals (RF line) are split into different frequency bands through a set of narrow band-pass filters (Healey et al., 1996). The filter outputs are envelope detected, weighted and compounded according to some rule in order to generate a 'speckle-reduced' image line (A-line). There are different compounding techniques, such as, for example, taking some average or the maximum of the envelopes of all the narrow band filter outputs. Figure 1 schematically illustrates a 1D RF signal, x (t), that, after 'frequency splitting', goes through an envelope detection and weighting stage and so is compounded back into a 'speckle-reduced' envelope, y(t).

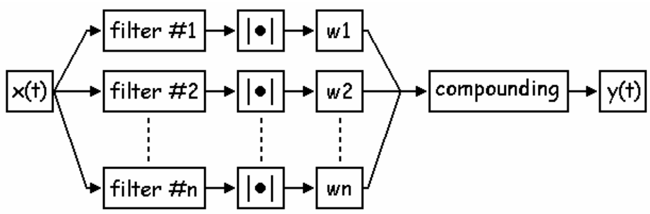

Figure 1 – 1D SSP basic scheme showing the input RF signal, x(t), the narrow-band splitting filters, the envelope detection and weighting stage, the compounding and its output – the 'speckle-reduced' envelope, y(t).

Even thought the speckle pattern 'crosses' the image from line to line, SSP is essentially a 1D method, with each single RF line being processed individually. After the last line was processed, all single lines can be combined to form a 'speckle-reduced' B-mode image. The number of filters changes with the application but in medical ultrasound imaging it is normally used 3 to 7 filters. Figure 2 shows the frequency spectra of a typical ultrasound pulse (bold dotted line) and its corresponding echo (dashed line). A truncated (at –40dB) Gaussian pulse was used, with a 3 MHz central frequency, 1.5 MHz bandwidth (measured at 50% of the peak) and 18 MHz sampling rate (the resultant pulse is 28 point long). The spectra of 5 Gaussian band-pass filters, with a 55% area overlap, are shown superimposed on the spectrum (solid lines).

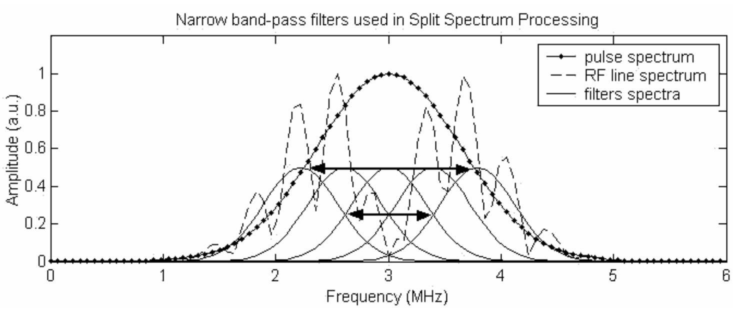

Figure 2 – Amplitude spectra of an ultrasound pulse and its associated RF line, with 5 narrow band-pass filters spectra used in the 1D SSP method.

The SSP tackles the speckle reduction problem assuming that the backscattered signals emanating from structures in the region of interest are shown consistently in all frequency bands (within the pulse bandwidth limits), whilst the speckle pattern (being more noise-like) will be erratically different. Relying on this idea, the compounding of all bands hopefully lead to a structure enhancement and speckle reduction. As can be seen on figure 2, even though the effective combined bandwidth of all the filters may be slightly wider that the original signal spectrum, each individual filter has a relatively narrow bandwidth, implying that the final compounded signal will be more time-stretched (lower resolution) than the original one. In this sense, there is, strictly speaking, a trade-off between the speckle reduction and resolution loss.

The angular compounding method does not suffer from this resolution loss limitation but it requires that the analysed structure be imaged from a number of different scanning directions. After that, all these images are compounded, generating an output with reduced speckle interference. The best results are obtained with a full 180° range - a severe practical limitation for medical imaging in particular. Another limitation relies on the fact that the final image is available only after the acquirement of the last image, meaning that this method cannot be used in real time.

We were aiming a speckle reduction method that could overcome the limitations of these two later methods: a single image one (unlike angular compounding) with no resolution loss (unlike SSP). As the SSP's bottleneck relies on its inherent resolution loss problem, we have overcome this limitation with the introduction of new modifications, by extending the conventional method to two dimensions, as schematically illustrated in figure 3. The 2D RF image, x (m,n), is split by a set of 2D wide-band filters. After the envelope detection stage, the filter outputs are weighted / normalised and compounded back into a 'speckle-reduced' image. The speckle decorrelation was achieved by using directive wide-band filters, applied at different directions in 2D frequency domain. It can be shown that each such filter outputs a different interference pattern (speckle) while leaving the structure relatively intact. Hence, a wide-band filtering at different directions is able to decorrelate the speckle while keeping the image's original resolution.

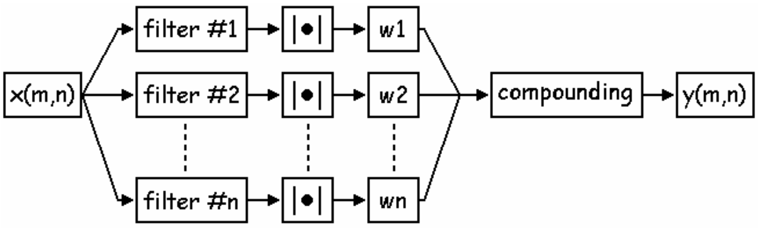

Figure 3 – 2D SPP basic scheme showing the input RF image, x(m,n), the wide-band filters, the envelope detection and weighting stage, the compounding and its output, the 'speckle-reduced' image envelope, y(m,n).

Radial Gaussian filters (truncated at –40dB) have been used, with examples shown in figure 4: a single filter on the left-hand side and a set of 7 superimposed filters on the right. Note that, even though there are some similarities between the two approaches, the use of directive filters makes the new technique a 'phase diversity' method, rather than a frequency diversity one (hence the designation: split phase processing, 'SPP'). The bandwidth of a single filter is chosen to be close to that of the ultrasound pulse, because the latter essentially determines the image bandwidth, and, hence, its resolution. The filter radius corresponds to the pulse central frequency, assuming the convention that, in 2D frequency space, the image's centre holds the low frequency components (image's brightness) and the edges, its high frequency components. The filters illustrated in figure 4 have a radius of 0.30 (assuming that 1 is the normalised distance from image centre to its border) and a relative bandwidth of 50%.

As the ultrasound pulse has a preferable direction, given by its carrier frequency, the filter in the pulse's direction (in the present case, the horizontal direction) will output the strongest result, whilst the ones at the orthogonal direction (any filter close to the vertical direction) will output the weakest results. Because of this reason, the normalisation becomes an important stage, providing the same 'strength' for all pre-images. They were normalised by the area, meaning that the sum of all pixel values for each image is equal to 1.

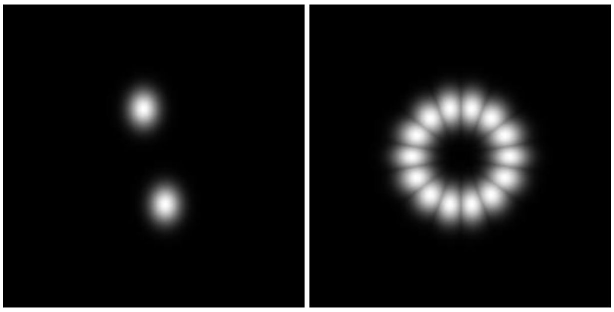

Figure 4 – Example of 2D directive radial filters used in the 2D SPP method. Left: one 2D directive Gaussian radial filter; right: superposition of 7 filters.

Probably, the most accurate way to analyse the performance of a speckle reduction method should be by visual inspection. However, in order to quantify the method's performance level, the most useful parameter is the SNR, that is calculated by the ratio between the mean (μ) and the standard deviation (σ) of an area known to have certain homogeneity (eq. 01).

$$SNR = \frac{\mu}{\sigma} \tag{1}$$

The SNR may vary inside an image, from region to region and the image's overall SNR should include statistics from the structure as well as from the background. With a image like figure 5, where there are two different regions (A and B), the resultant SNR and CNR, also known as pooled SNR and CNR, can be calculated via the following equations (Chen et al., 2003):

$$SNR = \frac{\mu_A + \mu_B}{2 \cdot \sqrt{\sigma_A^2 + \sigma_B^2}} \tag{2}$$

$$CNR = \frac{|\mu_A - \mu_B|}{\sqrt{\sigma_A^2 + \sigma_B^2}} \tag{3}$$

The phantom used to generate the B-mode image shown in figure 5 is made with point scatterers distributed with random position, magnitude and phase. The scatterers in region B have an average magnitude that is twice the one used for the ones in region A. In order to avoid border's effects and ambiguities in the interface between the two sides of the image, these two regions were defined far from the edges by a distance corresponding to the pulse length. Even though SNR and CNR are such widely used image quality parameters, they do not include any measurement of the resolution loss. The use of the traditional 1D SSP method with a strong speckle reduction level will cause a blurring effect, smoothing the edges and enhancing the SNR as well as CNR, depending on how the two regions are chosen. On the other hand, such a strong speckle reduction level can degrade the image's resolution, making it hard to detect and distinguish certain small structures. The SNR and CNR can be used to classify an image quantitatively as long as they are used with criterion.

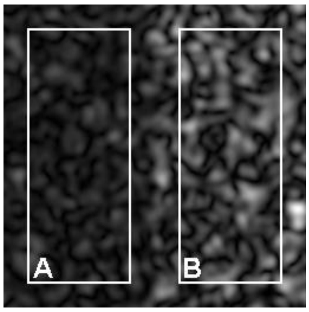

Figure 5 – B-mode image showing the regions of interest (A and B) where the SNR and CNR were calculated.

3. RESULTS

The outcomes of speckle reduction methods described above are presented for a simulated B-mode ultrasound image. Simulations were performed by convoluting arrangements of point scatterers (with random phase, magnitude and location) with a pulse consisting of a 2D carrier wave, propagating from left to right, with a Gaussian envelope, truncated at -40dB. The use of simulation is appropriate at this stage because it provides the possibility of calculating the speckle-free image: i.e., where the scatterers' phase may be changed directly in order to achieve only constructive interference. In this way, a reference image may be artificially constructed in order to evaluate the performance of the speckle reduction methods. The construction of such a 'golden standard' image is clearly not possible with real images, where the scattering structures, and their phases, are unknown. It is constructed by artificially constraining all the interference to be constructive, via the convolution of the pulse envelope with the magnitude of the point scatterers (Gehlbach, 1983).

Figure 6 shows the phantom used in the simulation, consisting of relatively densely packed point scatterers, with random phases and amplitudes, located randomly across the region to be imaged (note that for ease of display, only the magnitudes of the scatterers are displayed in an appropriate grey scale). It has two bright (mean scatterer magnitude high) circular structures embedded in a weaker background. A single scatterer (impulse) has been placed in an empty region at the top left-hand corner of the phantom in order to show clearly the broadening of the point spread function, (PSF = resolution loss) induced by the speckle reduction method. The background scatterers have a mean magnitude equal to one, whilst the left and right-hand side circles are formed with scatterers with mean magnitude equal to 3 and 2, respectively. In all the simulations, the scanning direction was from top to bottom, whilst the ultrasound pulse propagation direction was from left to right. The ultrasound pulse used has a Gaussian shape, with a 3MHz central frequency, and 50% bandwidth (absolute bandwidth of 1.5 MHz). A sampling frequency of 18 MHz was used throughout, with all images having dimensions of 256 x 512 pixels.

Figure 7 corresponds to the resultant B-mode image, which is the envelope of the RF image obtained via convolution of the ultrasound pulse with the scatterer set. Clearly, speckle is responsible for a noticeable degradation of image quality, introducing some difficulty in detecting edges. The fine 'worm-like' structures snaking through the image are a direct consequence of the interference effects that give rise to the speckle artefact pervading the entire image, wherever interference can occur.

The speckle-free image is shown in figure 8, obtained by convoluting the magnitude of the point scatterers with the pulse envelope, so that no interference is possible (Gehlbach, 1983). Contrary to expectation, the image brightness is nowhere uniform, even though the phantom structure is clearly evident. The single blob on the top left-hand corner has the same size as the one shown previously in figure 8, confirming (as expected) that there are no interference effects in that region. This image indicates the best that a speckle-reduction technique can achieve; any grey-scale variations are due to local fluctuations in the mean scatterer strength, and not interference.

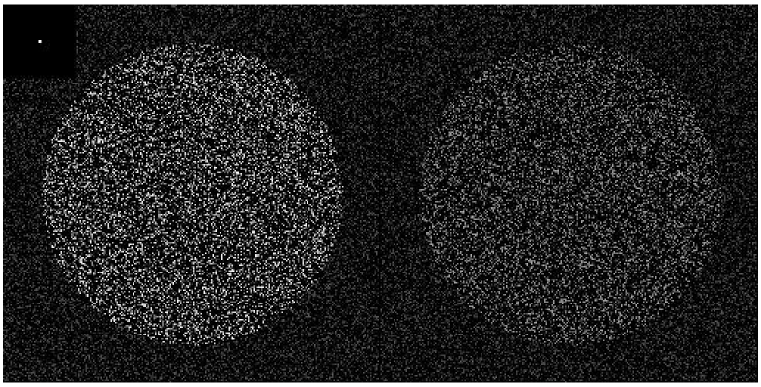

Figure 6 – Phantom used in the simulation (magnitude shown only).

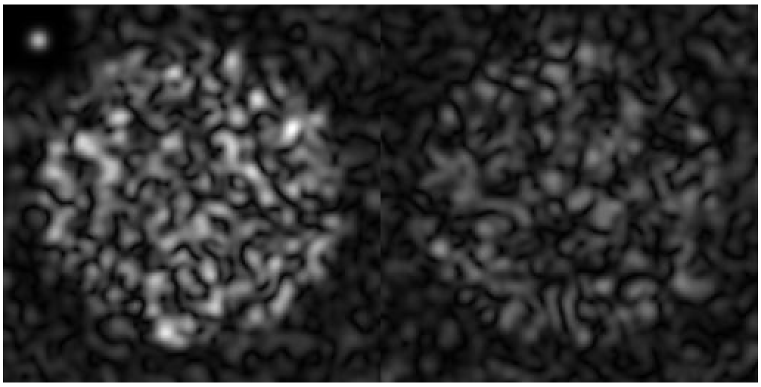

Figure 7 – Ultrasound B-mode image, showing strong interference presence (speckle).

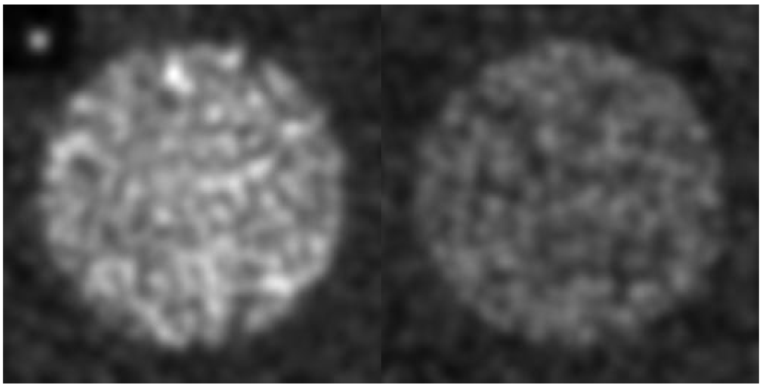

Figure 8 – Ultrasound speckle-free image (with constructive interference only).

Figure 9 shows the output of the conventional line-by-line SSP method, with five 1D 55% overlapping filters chosen and envelope average, as the compounding rule. Because of the choice of pulse, the individual A-lines making up the image run horizontally. The processing parameters were chosen empirically in order to achieve the 'best' result but the application will probably define the best way to manage with the trade-off between speckle reduction level X resolution loss. Some deep minima are still present, indicating that the speckle reduction is far from being optimal. More complete speckle reduction is achievable by choosing more filters, resulting, however, in even more resolution loss than that already evident (particularly from the size of the PSF).

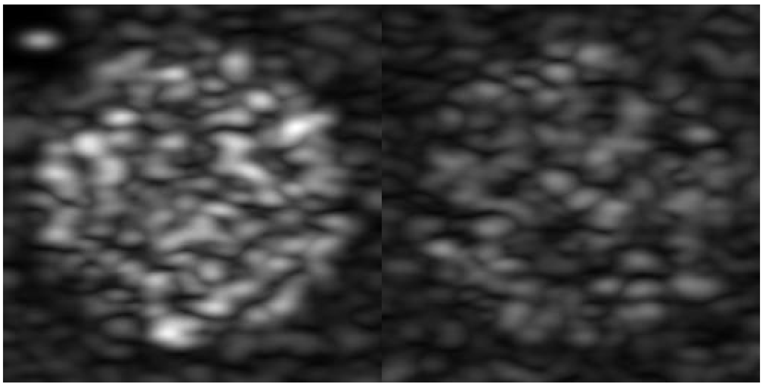

Figure 9 – Result of the traditional 1D SSP method.

Fig. 10 shows the output of an angular compounding method, representing the average of 6 images obtained at equal angular intervals over a 60° viewing range (which we regard as a practicable upper limit in most medical applications, although best results are obtained with a full 180° range).

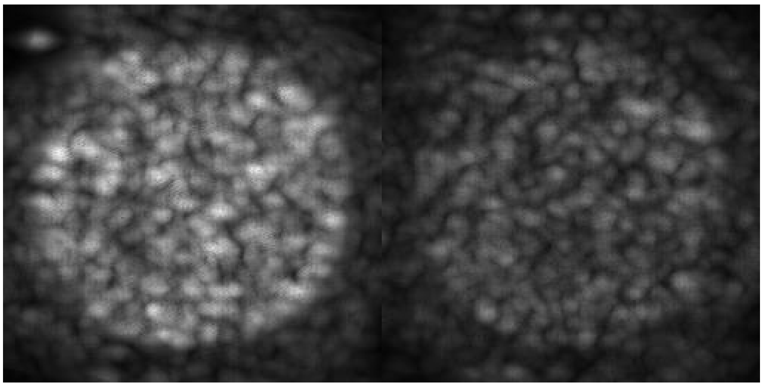

Figure 10 – Outputs of angular compounding over a 60° range

Figures 11 to 13 show, on the left-hand side, three (first, fourth and seventh) of seven wide-band directive filters used, whilst their outputs are shown on the right-hand side. These filters were designed as the ones shown in figure 4, i.e. a 0.30 radius and a 50% bandwidth. As can be seen, all these pre-images have basically the same structure shape but corrupted by speckle in different ways. Another important point is the single blob, present on the top left-hand corner of all outputs: it has the same symmetrical shape in every single output, indicating that they did not experience any noticeable resolution loss. As the other four missing filters (second, third, fifth and sixth) are very similar to these later ones, they are not shown.

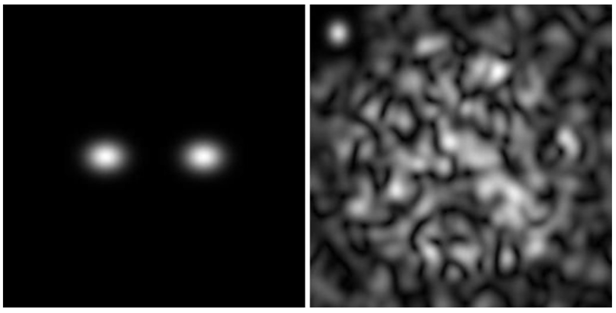

Figure 11 – Left: first directive filter, at 0°. Right: filter output example.

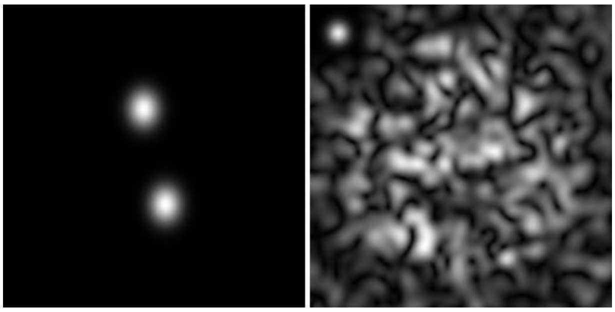

Figure 12 – Left: fourth directive filter, at 77.1°. Right: filter output example.

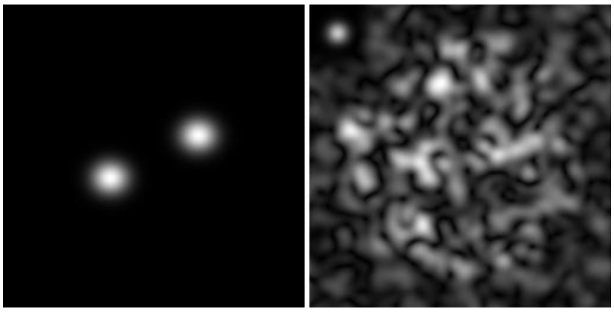

Figure 13 – Left: seventh directive filter, at 154.3°. Right: filter output example.

After the filtering stage and envelope detection, these pre-images are normalised to have the same ‘brightness’ (sum of all pixel values = 1) and compounded to produce the final output image. In this investigation, simple averaging was used; however, different compounding techniques, as well as another normalisation or weighting stage, are also possible.

Figure 14 shows the output of Split Phase Processing: it clearly represents an improvement over conventional SSP, both in terms of approximation to the speckle-free image, as well as retention of the original resolution. Also, the edges of the object appear to be sharper. In terms of speckle reduction level and resolution loss, this result is comparable to the angular compounding one, but is achieved with only a single image. A qualitative visual inspection suggests that the overall image appearance is not to unlike that of the speckle-free image.

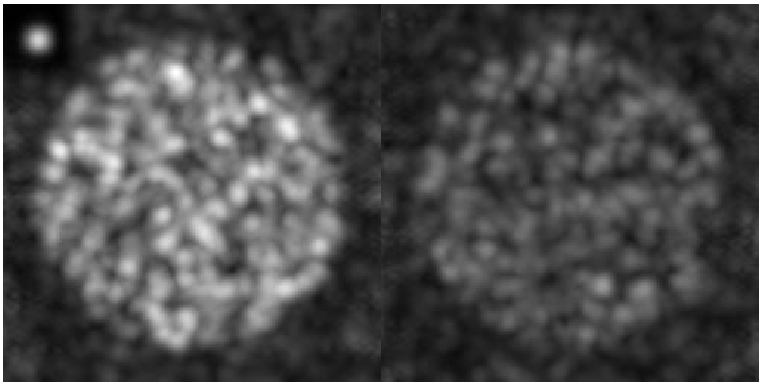

Figure 14 – Result of the new 2D SPP method with directive radial filters.

The performances of the different speckle reduction methods were quantitatively judged, in terms of SNR and CNR, with respect to a two-region phantom, with a sample B-mode image shown in figure 5. The two regions of interest (ROI) A and B have dimensions 216 x 88 pixels each, whilst the image size is 256 x 256 pixels. Simulation parameters were the same as those used for figure 7. The pooled SNR (eq. 02) and CNR (eq. 03) were calculated using the mean and standard deviation of regions of interest A and B for a 20 different image set, and their averages, with 1 standard deviation, are summarised in table 1, showing the improvement, relative to the original B-mode image, brought about by the methods described above. The numerical results appear to closely agree with the impressions of a visual inspection of the images.

Table 1 – Relative average SNR* and CNR* with 1 standard deviation of 20 sets of images (* relative to the B-mode image).

Image	SNR*	CNR*
Unprocessed B-mode	1	1
Speckle-free	3.35 ± 0.264	3.15 ± 0.311
Conventional SSP	1.32 ± 0.024	1.32 ± 0.029
Angular compounding	1.92 ± 0.107	1.73 ± 0.129
2D SPP	2.01 ± 0.072	1.95 ± 0.096

4. DISCUSSION AND CONCLUSION

The traditional 1D SSP method presents a relatively high-resolution loss limitation, inherent to the method, impinging a strictly trade-off between the speckle reduction level and the resolution loss. On the other hand, the new 2D SPP with directive filters does not present a noticeable resolution loss, voiding the conventional method's limitation. The fact that these filters have their spectra wider than the ultrasound pulse spectrum helps to keep the image's original resolution.

Even though the SSP (or frequency diversity) is a single-image method, it tackles the speckle reduction problem in a similar way as the frequency compounding (a multi-image method), where a set of images from the same structure is acquired by using ultrasound pulses with different frequencies. Likewise, the use of directive filters in the SPP method (also another single-image method) has a similarity with the angle compounding method (another multi-image method), where the images are acquired from different angles in order to decorrelate the speckle. However, the angle compounding method has a practical limitation due to the difficulty in acquiring images from certain angles, besides the fact that it is a multi-image method, whilst the SPP requires a single image only.

The quantitative analysis (table 1) agrees with visual inspection in the sense that the new SPP method is able to strongly perform speckle reduction while keeping edges details, without any noticeable resolution loss.

The filtering parameters, such as number of filter, bandwidth or superposition level, were determined empirically, in order to get the 'best' results. Otherwise, once these parameters were found, they can be kept unchanged and be used again for the same kind of ultrasound pulse, providing some robustness level to the method.

5. ACKNOWLEDGEMENTS

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Correspondence: e-mail: *grossi@ceb.unicamp.br; address: UNICAMP/CEB, caixa postal 6040, Campinas/SP, Brazil, 13084-971; telephone: +55 19 3788-9291 / 9260; fax: +55 19 3289-3346